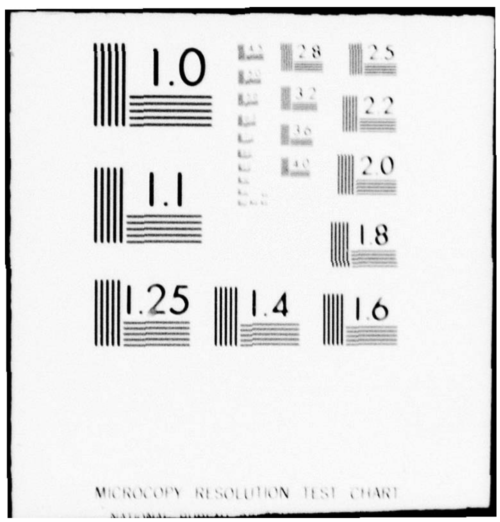
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COMPUTERIZED MEASUREMENT OF VENTILATION WITH FOUR CHEST WALL MAGNETOMETERS

C.H. Robertson, Jr., M.E. Bradley, L.M. Fraser, and L.D. Homer

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W. F. Miner, CAPT, MC, USN
Commanding Officer
Naval Medical Research Institute

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C.H. ROBERTSON, JR., M.E. BRADLEY, L.M. FRASER

and L.D. HOMER

Hyperbaric Medicine and Physiology Department

Naval Medical Research Institute

Bethesda, Maryland

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ABSTRACT

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Index Terms

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INTRODUCTION

In many experimental situations spirometers and flow meters cannot be satisfactorily employed to measure ventilation because the devices themselves alter respiration. Simply breathing through a mouthpiece affects the respiratory pattern (5). Most of the apparatuses add dead space and resistance to breathing, which alter ventilation. Such equipment also limits the freedom of movement of the subject and thus affects the ventilatory response to exercise.

In an attempt to overcome these difficulties we have turned to the use of pairs of electromagnets and magnetometers placed on the chest wall to measure linear dimension changes and estimate lung volume changes as previously described by Mead and coworkers (7, 8). However, we soon became aware of the fact recognized by previous investigators working with magnetometers (7, 9, 10) that single anteroposterior pairs of magnetometers on the rib cage and abdomen that they used, while fairly quantitative for normal tidal volumes, became increasingly inaccurate as breath volume approached the vital capacity. During resistive breathing or forced expirations the two pairs also were not sufficient as observed previously by others (1, 2, 9, 10). As the volume is increased from normal or during increased respiratory efforts, the rib cage and abdomen no longer behave with single degrees of freedom, so single magnetometer pairs cannot quantitatively measure these maneuvers. That is, there are changes in the cross-sectional dimensions of the chest wall in the lateral dimension which are not reflected in anteroposterior (A-P) movements (Figure 1).

Therefore, we have used four pairs of magnetometers placed A-P and laterally across the rib cage and abdomen to include these additional degrees of freedom of movement of the chest wall. We developed a model of

chest wall movement which allows inclusion of these additional two lateral magnetometer signals and used a computer to correlate simultaneous volume and chest wall magnetometer signals in a manner similar to that described previously by Stagg, Goldman and Davis (11). However, because of the use of four magnetometer signals in our model instead of the two A-P magnetometers they used, we were unable to utilize their "isovolume comparison" calibration algorithm and turned to a least squares regression to determine the constants necessary in our model to correlate volume and magnetometer signals. This report describes the model used to include lateral as well as A-P movements of the rib cage and abdomen and presents data on the quantitative accuracy of our computerized method for measuring lung volume changes up to and including vital capacity maneuvers, during respiratory loading, and in various body positions.

METHODS

Six healthy adult men were studied (Table 1). One subject was a smoker. None had overt pulmonary functional abnormalities. In an attempt to determine the applicability of the model to different body habitus, subjects were selected whose height/weight relationships varied considerably and whose resting chest wall dimensions were quite different. All but one were naive to the respiratory measurements being performed. Electromagnet and magnetometer pairs were attached to the chest wall with adhesive discs in the midline both A-P and laterally at the level of the nipples to measure rib cage movements and at the level 2-3 cm above the umbilicus to measure abdomen-diaphragm movements. Subjects were seated in a volume plethysmograph breathing to the outside. Care was taken to insure that the electromagnet

and sensor in each pair were placed parallel to each other because if the angular relationship changes with respiration, errors in distance measurement will occur as the magnetic flux at the sensor will change.

Thoracic gas volume change from the plethysmograph and changes in outputs from the magnetometers were recorded simultaneously on FM tape for later computer analysis. Volume and magnetometer outputs were all zero suppressed to give no voltage signal at end-expiration during quiet breathing. Prior to each study the FM tape was calibrated by addition of six liters to the closed plethysmograph and by movement of each magnetometer ±2 cm in a plexiglass rack with machined blocks accurate to .01 cm. In preliminary studies each of the magnetometer pairs was found to have linear voltage change with distance change over the range 13-40 cm. For each subject recordings were made during quiet breathing (QB), a vital capacity manuever (VC), and during hyperpnea produced by adding dead space sufficient to produce tidal volumes of approximately 1.5 liters. Two subjects performed forced vital capacity maneuvers and breathed against inspiratory and expiratory resistances.

In addition, one subject was studied outside the plethysmograph breathing through a flexible hose into the closed box. This enabled us to make measurements in various body positions and during walking so that we could determine the suitability of the model during these different conditions.

In order to incorporate degrees of freedom to movement in both A-P and lateral dimensions, we have assumed that the rib cage and abdomen approximate cylinders whose base is an ellipse (Figure 2). Since the area of an ellipse is:

$$A = \text{radius}_1 \times \text{radius}_2 \times \pi \tag{1}$$

the cross-sectional area of the rib cage can be determined in the plane of the magnetometers as:

$$A_{RC} = D_{AP} \times D_{LAT} \times \pi/4 \qquad (2)$$

where: D_{AP} and D_{LAT} are AP and lateral diameters respectively. Since the magnetometers are zero suppressed at end-expiration.

$$A_{RC} = (M_1 + D_1)(M_2 + D_2)\pi/4 \qquad (3)$$

where: M_1 is the signal from the A-P magnetometer pair and D_1 is the distance between the A-P pair at end-expiration, which can be measured with calipers. M_2 and D_2 are the same for the lateral pair.

By similar analysis:

$$A_{AB} = (M_3 + D_3)(M_4 + D_4)\pi/4 \qquad (4)$$

thus, the cross-sectional areas of the rib cage and abdomen at any time are defined by the signals from the four pairs of magnetometers and the end-expiratory dimensions. The volume of these cylinders then would be:

$$V_{RC} = A_{RC} \times h_{RC} \qquad (5)$$

where h_{RC} = height of the rib cage cylinder

and

$$V_{AB} = A_{AB} \times h_{AB} \qquad (6)$$

where h_{AB} = height of the abdominal cylinder.

Since the diaphragm moves during respiration, the heights are not constant and the heights are obviously not measured in this system. However, we can reasonably assume that the sum of the heights is relatively constant as the top of the rib cage is attached to the pelvis by the relatively fixed vertebral column. Thus, we assume that:

$$K_1 = h_T = h_{RC} + h_{AB} \qquad (7)$$

then: $$h_{RC} = K_1 - h_{AB} \tag{8}$$

and from (5) and (8):

$$V_{RC} = A_{RC} \times (K_1 - h_{AB}) \tag{9}$$

also from (6):

$$h_{AB} = V_{AB}/A_{AB}. \tag{9a}$$

So:

$$V_{RC} = A_{RC} \times (K_1 - V_{AB}/A_{AB}) \tag{10}$$

If we also assume that the volume of the abdomen is constant (since, if one ignores the small amount of gas in the bowel, it is almost entirely filled with incompressible structures), then while area and height may change reciprocally,

$$V_{AB} = K_2 \tag{11}$$

and from 10 and 11,

$$V_{RC} = A_{RC} \times (K_1 - K_2/A_{AB}) \tag{12}$$

Finally, in order to compare predicted volume to the plethysmographic volume which is set to zero at end-expiration,

$$V_{RC} = V + K_3, \tag{13}$$

where K_3 is the volume inside the elliptical cylinder approximated by the rib cage at end-expiration. Thus, during respiration the change in volume of the rib cage cylinder which contains the lungs can be approximated by the equation:

$$V = K_1 A_{RC} - K_2(A_{RC}/A_{AB}) - K_3 \tag{14}$$

Since the change in lung volume that occurs with respiration is due partially to increasing rib cage dimension and partially to descent of the diaphragm and thus increases in abdominal diameter, the computer was also programmed so that given the constants in equation 14 it could

calculate the change in volume due to rib cage expansion (ΔV_{RC}) as

$$\Delta V_{RC} = \int h_{RC} \; \delta A_{RC} \tag{15}$$

and the change in volume due to descent of the diaphragm (ΔV_{Di}) as

$$\Delta V_{Di} = \int A_{RC} \; \delta h_{RC} \tag{16}$$

which from equation 12 can be seen to be

$$\Delta V_{RC} = \int [(K_1 - K_2)/A_{AB}] \; \delta A_{RC} \tag{17}$$

and

$$\Delta V_{Di} = \int A_{RC} \; \delta(K_1 - K_2/A_{AB}) \tag{18}$$

The recorded volume and magnetometer voltage signals were low pass filtered with an active quadripole filter (3 dB at 10 Hz) and analog to digital conversion performed at 25 samples/second by a PDP 11/40 computer (Digital Equipment Company, Maynard, Mass.). The computer was programmed to perform a least squares analysis to determine the values of K_1, K_2, and K_3 which produced the best fit of the volume and magnetometer data to the double elliptical cylinder model according to equation 14. In order to assess the degree to which the model fit the data, the computer also determined the correlation coefficient (R) and the variance around the line of identity for the data set. Once the computer had selected the appropriate constants for one type of respiratory maneuver, these could be specified for other tidal volumes, etc., to determine the predictive capacity of the model. In practice, it was found that while quiet breathing was acceptable for determination of the constants as in the two magnetometer method of Stagg, Goldman and Davis (11), slightly better estimates were generated if the signal in the four magnetometer channels was increased by increasing the tidal volume to about 1.5 liters either voluntarily or by adding dead space to the external airway. In the studies reported here,

the least squares analysis to determine the constants in equation 14 was performed during breathing through added dead space tubing. The constants so generated were used to assess the predictive value of the magnetometer signals to determine lung volume change during quiet breathing and a vital capacity in all subjects, during forced vital capacities and resistive breathing in two subjects, and during isovolume and Valsalva maneuvers in the one trained subject. In the studies outside the box, parameters generated while standing motionless were used to predict volume while walking and rocking from side to side erect, while sitting, while supine and while in the left lateral decubitus position.

RESULTS AND DISCUSSION

Initial Fit of Model

Calibration was performed on breaths of approximately 1.5 liters produced by rebreathing in order to increase the signal-to-noise ratio. Using the equation relating the chest wall to two elliptical cylinders and least squares analysis on the computer, in every case we were able to determine values of the constants which produced excellent correlations between actual volume change and that calculated from the magnetometer signals. An example of the close correlation is shown in Figure 3. Table 2 gives the numeric results in all of the subjects. Although there was a very high correlation between estimates of parameters (almost always .99 or higher), the percentage variation in parameter estimates (an index of the computer's ability to specify a parameter exactly) was low (highly specific), usually less than 2% of the parameter estimate.

However, with three parameters being used to perform a least squares fit, it is possible that a good fit would be obtained even with an erroneous model. Thus, Stagg, Goldman and Davis's computerized method (11) and the

earlier graphical "isovolume" methods (6-10) obtained good results using only antero-posterior dimension measurements to estimate normal and moderately increased tidal volumes, even though there are clearly other degrees of freedom of chest wall movement in the lateral dimension which that model cannot take into account (Fig. 1). One way to assess whether the current model is reasonable is to see if the estimated parameters correlate well with known anatomic values. Thus, one of the parameters estimated, K_1, is the total height of the rib cage plus abdomen. In Table 2, a list is given of the K_1 values and the height of the chest wall from sternal notch to symphysis pubis in these subjects. The predictions appear qualitatively correct and show a good quantitative correlation ($p > .92$ by paired T test). The other parameters, K_2 and K_3, do not have any exactly measureable anatomic correlates, so one can only say that in a qualitative way they are reasonable, with K_2 (abdominal cylinder volume) values ranging from 6 to 14 liters and K_3 (rib cage cylinder volume at FRC) values from 14 to 26 liters. The rib cage cylinder volume is larger than lung volume at FRC because it contains all the structures inside the skin surface where the magnetometers are placed, not just the lungs. Also, these values appeared to trend as one would expect in the different subjects; the smallest K_2 (6 liters) was in the most ectomorphic and the largest K_2 (14 liters) in the most endomorphic subject. Thus, it would appear that this model provides a reasonable approximation of chest wall movements during respiration.

Direct comparison of the ability of our four-channel model to correlate magnetometer signals and lung volume with the capacity of the two-channel computerized method of Stagg, et. al. (11) is difficult as their data are expressed differently from ours. Their primary interest was the measurement

of tidal ventilation and times of inspiration and expiration in subjects who were kept in a fixed supine position. Under these conditions, it appears that their method can accurately measure normal and moderately elevated tidal volumes. While we were concerned with the precise measurement of tidal volume, we were also concerned with the accurate prediction throughout the respiratory cycle of large volumes, such as the vital capacity, and forced respiratory maneuvers where two channel methods have difficulty (1, 2, 7, 9, 10). Moreover, we were interested in determining whether any magnetometer method can be used to accurately measure ventilation when a subject is allowed considerable freedom of movement.

Predictive Value of Model

The possibility remained that although ours was a reasonable model at rebreathing tidal volumes around 1.5 liters, it might be inapplicable to resting tidal volumes or very large tidal volumes which approach vital capacity. In order to assess this possibility we used the parameters determined during the calibration runs in each subject and compared their estimate of thoracic gas volume change with the plethysmographic value during quiet breathing and a vital capacity. Representative tracings for quiet breathing and a vital capacity in one subject are shown in Figure 4A and 4B respectively. The correlation coefficients and variances for all subjects are given in Table 3.

For quiet breathing the parameters generated during rebreathing had excellent predictive value in all subjects with R values .95 or higher and variances of 2.5 milliliters or less (.2 to .5% of the tidal volume being measured). Over the full range of the vital capacity the

close correlation persisted, with R values .97 or higher. Variances around the line of identity were higher (14-121 milliliters) but still only an average of 1.2% of the volume measured. While some of the increased variance would be expected on the basis of the larger volume of the breaths, at least four other factors seemed to play a role. First, since the subject was seated inside a plethysmograph, there was some drift of the volume signal with time due to heating and humidifying the air in the box. Thus, over the 15-20 seconds of a slow vital capacity maneuver some deviation of the actual and predicted volume signals would be expected and was suggested by the plethysmograph volume signal not returning to exactly the same FRC after the vital capacity maneuver. Secondly, there was a small drift in the magnetometers and their electronics which contributed to a lesser extent. A third reason for the increased variance was that at the extremes of vital capacity, some subjects showed a clear deviation of the predicted volume signal from the line of identity. In the subject illustrated in Figure 4B this occured near TLC, while in other subjects this was observed near RV. It may be that the magnetometers and electromagnets change their angular relationship to each other as one gets to the extremes of lung volumes and thus artifactually measure increased or decreased magnetic flux and thus distance. This would seem particularly likely in the lateral dimension of the rib cage where the two sides of the chest wall may significantly change their angular relationship with large lung volume changes. It is also likely that at the extremes of lung volumes the model may begin to break down. For example, the model assumes that the total height of the rib cage and abdominal cylinders does not change because the top of the rib cage is connected to the bottom of the abdomen by the vertebral column. However,

as one approaches TLC there is a tendency to straighten the spine to increase rib cage volume and as one approaches RV to bend the spine to decrease volume. To the extent that this occurs, the model would be expected to be inaccurate.

A fourth reason for the larger variances during vital capacities is also illustrated in the example given in Figure 4B, and to some degree is seen at all tidal volume ranges including quiet breathing (Figure 4A). That reason is that there tends to be looping of the estimated volume around the line of identity with low estimates during inspiration and high estimates during expiration. During inspiration, in addition to air coming into the chest from the outside, blood shifts from the extremities into the thorax, while the reverse occurs during the expiration. The thoracic volume measured by the magnetometer will include the blood shifts but the plethysmograph will not, as the whole body is inside the plethysmograph. The computer estimation of parameters partially compensates for this by adjusting parameter values to the best fit, but that fit will work only if the amount of blood shifted remains proportionally the same throughout the vital capacity maneuver which is unlikely. The fit also certainly will not work if there is a lag in blood shifts into the thorax during inspiration and back out during expiration. These lags would cause the magnetometers to underestimate volume during inspiration and overestimate during expiration, just as observed in these subjects. This looping is clearly the most important cause of inaccuracy of the model's prediction of ventilation. This is especially true at tidal volumes significantly less than the vital capacity and is unavoidable if thoracic volume change is used to measure ventilation. The extent to which this

blood shift can cause error in estimated lung volume is illustrated by the effect of a Valsalva maneuver (Fig. 5). After a normal quiet breath the subject generated a maximal expiratory effort against a closed glottis. There was a small decrease in both plethysmographic and magnetometer volume and then magnetometer volume decreased by approximately an additional 400 ml. This volume of blood shift is slightly larger than Bahnson's estimate of a blood volume shift of 147 - 306 ml with a Valsalva (3). The reason for this is probably because the magnetometers will also measure the volume of blood which is squeezed out of the abdomen into the extremities. If these speculations are correct it may be possible to use the difference between magnetometer volume and plethysmographic volume to measure blood shifts under a variety of physiologic conditions. A similar conclusion has been reached on the basis of unpublished observations by J. Vorosmarti and M. Goldman (personal communication).

<u>Fractionation of Rib Cage and Abdominal Volume</u>

Since this model assumes that the lungs exist inside an elliptical cylinder whose volume can be changed either in cross-sectional area (which itself has two degrees of freedom, A-P and lateral) or in height, the computer program contained the capacity to calculate the volume contributed by height change, $\int A dh$, and that contributed by cross-sectional area change, $\int h dA$. It was assumed that these differentials should correspond to the volume moved by the diaphragm-abdominal muscles and rib cage muscles respectively. To test this hypothesis the one subject experienced in respiratory maneuvers performed the "isovolume maneuver" used in earlier two mangetometer systems for calibration (6-8, 10). In this maneuver the subject attempts to shift between expansion of the abdomen and expansion of the rib cage while maintaining a fixed lung volume and in such a way that minimal pressure is exerted on the gas in the lungs. Figure 6A shows the actual volume signal from the

plethysmograph, the total volume determined by the magnetometers, and the volumes contributed by the rib cage and abdomen during the "isovolume maneuver" in this subject. It is apparent that, despite wide fluctuations in rib cage and abdominal volumes, the volume prediction is close to the actual volume during this maneuver. Figure 6B is a plot in the form usually used for calibration in earlier two magnetometer studies with rib cage volume plotted against abdominal volume. It is apparent that with the four magnetometer method we have used these volumes have the appropriate negative correlation with a slope of approximately -1.0 and that the two volumes do not exhibit the slightly curvilinear relationship which is often seen when the two magnetometer system is used (6, 7,10). It would appear that our model has the capacity to accurately measure the fractionation of ventilation between rib cage and abdomen-diaphragm movements but that it does not require the use of the awkward "isovolume maneuver" which demands a highly trained subject. Thus, our method shares with the two channel computerized method of Stagg, Goldman and Davis (11) the advantage that it is easier to utilize in naive subjects with the obvious benefits that different subjects may be used and the results are less likely to be colored by the tendency of trained subjects to be more aware of their ventilatory pattern, which awareness may affect the parameter being measured (5).

Figure 7A shows an example of the fractionation of volume moved by the rib cage and abdomen-diaphragm during quiet breathing in one of our subjects. The finding in these subjects sitting upright that 74% (range 47-91%) of the volume change was accomplished by the rib cage and 26% by the abdomen-diaphragm is similar to previous estimates of the fractionation by other techniques. Thus, Bergofsky observed by a plethysmographic technique that the abdomen contributes 26% of the volume increase during quiet breathing in the upright

position (4). Using the two magnetometer method Grimby, Bunn, and Mead estimated that the abdomen accounts for 24% of the volume in seated subjects (6); Sharp and associates attributed 28% of the tidal volume to abdomen displacement in young men, 39% in young women, and 26% in older men (9); and Konno and Mead also estimated 28% (7).

Figure 7B illustrates the fractionation of volume moved by the rib cage and abdomen-diaphragm during a vital capacity maneuver in one of the subjects. In these six subjects the fractionation of a vital capacity was similar to that observed during quiet breathing; 22% of the volume was accomplished by the abdomen-diaphragm and 78% (range 60-91%) by the rib cage. This result is similar to the estimate of Konno and Mead using a two magnetometer method that 70% of the vital capacity is accomplished by the rib cage (7). Over the range of the inspiratory capacity, Sharp and associates observed that "almost all" of the volume was accomplished by the rib cage (9). Our result conflicts, however, with that of Agostoni and co-workers, who found using a geometric technique that over the vital capacity the rib cage moved only 37% of the volume in the sitting position (2). Our four magnetometer method of fractionating volume between the rib cage and abdomen shares with the two magnetometer methods the advantage that, compared to geometrical (1, 2) or plethysmographic methods (4), no fixed assumption about the position of the partition between rib cage and abdomen must be made; and in our model it is assumed that the site of partition must move with breathing just as the diaphragm does. This may explain the different result of Agostoni's group.

<u>Increased Respiratory Effort</u>

To assess the utility of this method for measuring lung volume changes during more strenuous respiratory efforts than unloaded quiet breathing, slow vital capacities or rebreathing, two subjects performed forced vital

capacity maneuvers and breathed against external expiratory and inspiratory resistances while seated in the plethysmograph. This comparison seemed particularly crucial in light of the recent observations of Melissinos and associates (9) demonstrating that there are changes in the configuration of the chest wall during forced expiration which make only A-P dimension changes incapable of measuring the volume change accurately. Thus, the forced vital capacity and other large respiratory efforts represent a test of whether the four channel magnetometer model proposed here offers the benefit of being usable under conditions where a similarly computerized two channel system would falter.

Using the calibration parameters generated during rebreathing, there was good correlation between plethysmographic volume change and the volume predicted by the four channel magnetometers during the forced vital capacity maneuver (R = .99 and .98 with variances of 68 and 100 ml in the two subjects). An example of the correlation is shown in Figure 8A. The expiratory and inspiratory resistances were also well predicted (R = .99 and .98 with variances of 1.7 and 12.3 ml for expiratory resistance, and R = .98 and .93 with variances of 3.7 and 29.7 ml for inspiratory resistance).

The fractionation of the forced vital capacity between rib cage and abdomen is illustrated in Figure 8B. The relative amount contributed by the abdomen or rib cage was essentially the same as that observed for the slow vital capacity maneuvers in these two subjects.

The forced vital capacities exhibited a larger variance than the slow vital capacities associated with wider inspiratory-expiratory looping in Figure 8A and a continued decrease in magnetometer volume late in the forced expiratory effort when expired volume from the plethysmograph had plateaued in Figure 8B. This may be due to the larger intrathoracic

and intra-abdominal pressure swings during the forced maneuver and thus larger shifts of blood into and out of the thoracoabdominal cavity. This is thus similar to the suggested reason for the smaller looping observed during less strenuous efforts (quiet breathing, rebreathing, slow vital capacity) and is equivalent to the blood shift discussed above in relation to the Valsalva maneuver.

<u>Effect of Posture and Movement</u>

If this method of determing ventilatory volume and fractionating it between rib cage and diaphragm-abdomen is to have a major role in physiologic monitoring, it must be relatively unaffected by posture or movement. In order to assess whether this technique was sufficiently accurate under those various conditions to be usable, we performed a series of measurements on one subject who was outside the box but breathing into it through a flexible tube. This allowed movement and changes in posture. Calibration of the magnetometers was performed with the subject rebreathing from the plethysmograph standing erect and motionless. The parameters so generated were used to predict ventilatory volume while rocking from side to side erect, walking in-situ, sitting, lying supine, and lying in the left lateral decubitus position.

During rocking, the model predicted volume quite accurately (R = .98, variance 4.1 ml). During walking (Figure 9) the R value remained fairly good (.97), but the variance rose to 17.7 ml. The step artifacts can readily be observed in the tracings and could be corrected for by extrapolation or averaging techniques if the magnetometers were to be used for exercise monitoring. The high correlation persisted for the sitting (R = .98, variance 9.1 ml) and lying supine positions (R = .99, variance .9 ml), but decreased slightly in the decubitus position (R = .95, variance 15.9 ml).

However, the end-expiratory chest wall dimensions (to which the magnetometer signals are zero suppressed) changed significantly when the subject assumed the decubitus position, whereas the changes in end-expiratory chest wall dimensions were small in the other positions. When the new end-expiratory dimensions were substituted in the decubitus position while keeping the values of the constants of equation 14 constant at the values predicted in the standing position, the prediction was greatly improved (R = .99, variance 1.7 ml) and even vital capacity was accurately predicted (R = .99, variance 32.8 ml).

It would appear that the model proposed here, while not perfect, is sufficiently accurate to allow it to be used as a resistanceless method of monitoring ventilation under a wide variety of tidal volumes, postures and exercise states as long as the chest wall dimensions at FRC are not dramatically altered. Since it does not require a mouthpiece, does not impose a totally fixed body position, does not add any load to respiration, can be used even when respiratory efforts are great, and can be used in naive subjects, it may be particularly suited to studies of respiratory control mechanisms and has a potential role in monitoring of critically ill patients.

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FIGURE LEGENDS

Figure 1. The changes in A-P and lateral dimension of the rib cage and abdomen in centimeters are shown for a vital capacity maneuver. There are both amplitude and phase differences between A-P and lateral dimensions for both the rib cage and abdomen; that is, there are degrees of freedom for lateral chest wall movements independant of A-P movements. The four magnetometer model was designed to include all of these degrees of freedom of chest wall movements in an attempt to improve the accuracy of magnetometer methods of measuring ventilation and fractionating it between rib cage and abdomen.

Figure 2. The elliptical cylinder model used assumed that changes in distance measured between magnetometers placed as shown by the arrows in this diagram would define the changes in cross-sectional area of the elliptical cylinders approximating the rib cage and abdomen.

Figure 3. This plot of the volume change measured by the magnetometers against the plethysmographic volume change during rebreathing in one of our subjects is representative of the close correlation between the actual and estimated volume changes attained during calibration with the elliptical cylinder model.

Figure 4. A. This tracing demonstrates for a quiet breath the correlation obtained in a representative subject between plethysmographic volume change and volume change predicted by the magnetometers using the calibration parameters determined during larger tidal volumes induced by rebreathing.

Figure 4. B. This tracing of a vital capacity maneuver in the same subject illustrates that a good correlation also results at volumes larger than those used for calibration.

Figure 5. During a Valsalva maneuver at the end of a quiet breath, both plethysmographic and magnetometer volumes decrease slightly due to compression of gas. Then the magnetometer volume decreases a furthur approximately 400 cc as blood is shifted from the thorax and abdomen into the extremities.

Figure 6. A. Plethysmographic volume change, total volume change determined by the magnetometers, and volume changes calculated for the rib cage and abdomen-diaphragm are plotted against time for an "isovolume maneuver".

Figure 6. B. This plot of rib cage against abdominal volume change demonstrates that the appropriate reciprocal relationship exists.

Figure 7. Examples of the fractionation of ventilation between rib cage and abdomen-diaphragm volume changes are shown for quiet brething (A) and a vital capacity (B).

Figure 8. A. The correlation of magnetometer and plethysmographic volume changes are shown during a forced expiratory vital capacity and the subsequent inspiration.

Figure 8. B. This plot of plethysmographic volume change, total magnetometer volume and volume changes for the rib cage and abdomen during the same forced expiratory vital capacity as 8A illustrate the similar abdomen-rib cage fractionation as in the slow vital capacities. The continued decrement in total magnetometer volume after plethysmographic volume has plateaued may be due to shifts of blood out of the thorax.

Figure 9. The capacity of the magnetometers to measure ventilation despite movements and position changes are illustrated by these tracings of volumes vs time (A) and magnetometer volume vs plethysmographic volume (B) during walking.

TABLE 1

PHYSICAL CHARACTERISTICS OF SUBJECTS

SUBJECT	AGE	HEIGHT	WEIGHT	BODY HABITUS	RIB CAGE D_{AP}	RIB CAGE D_{LAT}	ABDOMEN D_{AP}	ABDOMEN D_{LAT}
1	31	180 cm	70 kg	ecto	20.4 cm	33.2 cm	18.1 cm	29.2 cm
2	25	185 cm	84 kg	meso	20.7 cm	36.0 cm	20.7 cm	30.5 cm
3	24	178 cm	86 kg	endo	23.8 cm	35.8 cm	26.8 cm	33.3 cm
4	25	183 cm	68 kg	ecto	16.6 cm	26.8 cm	17.3 cm	26.3 cm
5	28	178 cm	84 kg	meso	21.3 cm	36.2 cm	26.4 cm	32.9 cm
6	54	185 cm	97 kg	endo	29.6 cm	40.4 cm	29.5 cm	35.2 cm

TABLE 2

FIT OF INITIAL CALIBRATION

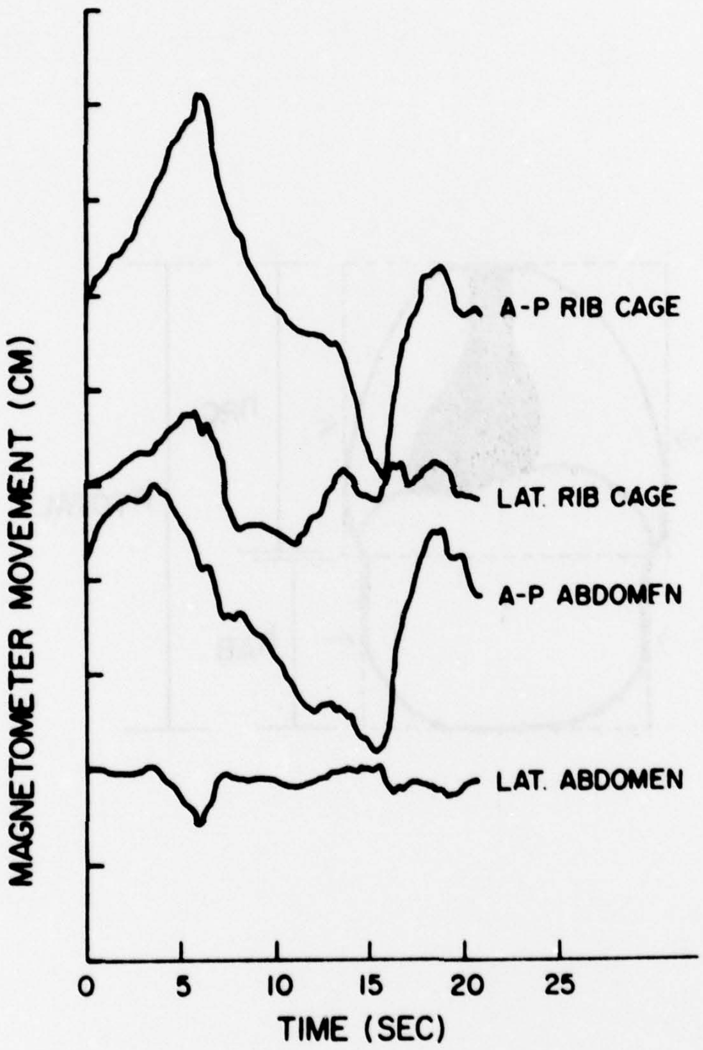
Subject	R Value	Variance	K_1	Chest Wall Height*
1	.997	2.17 ml	48 cm	48 cm
2	.999	.64 ml	55 cm	57 cm
3	.995	1.39 ml	55 cm	56 cm
4	.994	2.14 ml	63 cm	55 cm
5	.996	1.59 ml	54 cm	56 cm
6	.999	1.44 ml	57 cm	59 cm
Mean	.997	1.56 ml	55 cm	55 cm

* Height from sternal notch to symphysis pubis.

TABLE 3

PREDICTIVE CAPACITY OF MODEL

Subject	Quiet Breathing R Value	Quiet Breathing Variance	Vital Capacity R Value	Vital Capacity Variance
1	.99	.9 ml	.98	45 ml
2	.96	2.4 ml	.98	121 ml
3	.98	1.4 ml	.97	89 ml
4	.99	1.1 ml	.99	14 ml
5	.95	2.5 ml	.99	36 ml
6	.99	1.7 ml	.98	58 ml
Mean	.98	1.7 ml	.98	61 ml



MAGNETOMETER MOVEMENT (CM)
A-P RIB CAGE
LAT. RIB CAGE
A-P ABDOMEN
LAT. ABDOMEN
0
5
10
15
20
25
TIME (SEC)

Figure 1.

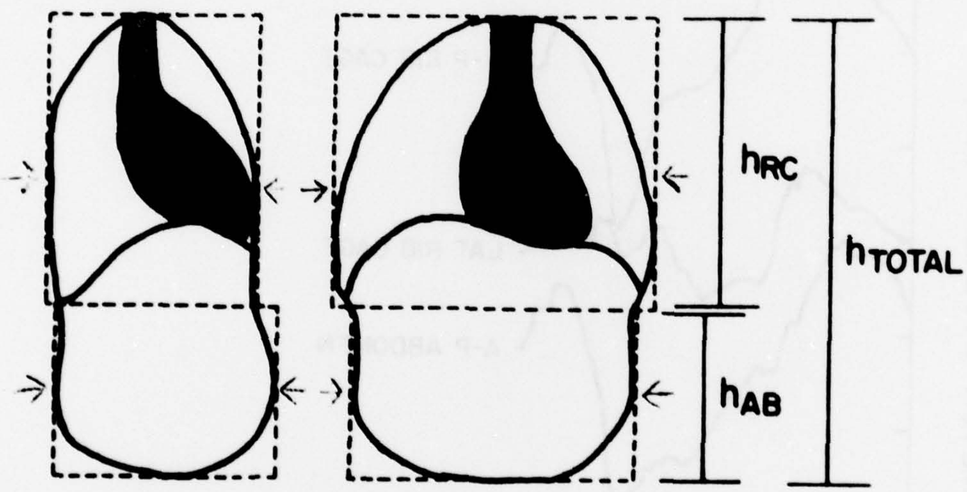
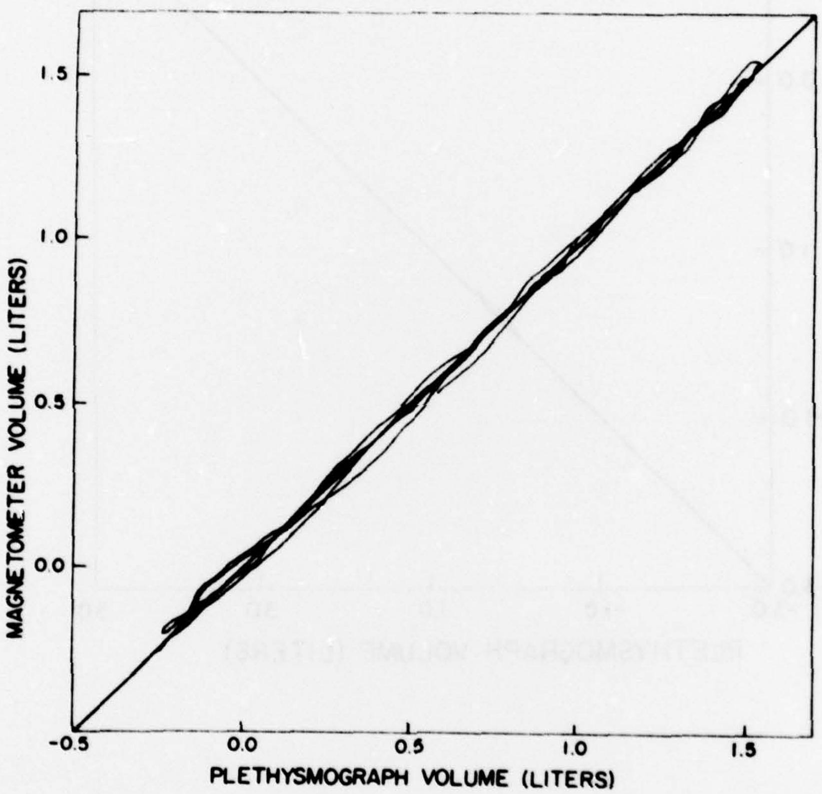


Figure 2.



MAGNETOMETER VOLUME (LITERS)
1.5
1.0
0.5
0.0
-0.5
0.0
0.5
1.0
1.5
PLETHYSMOGRAPH VOLUME (LITERS)

Figure 3.

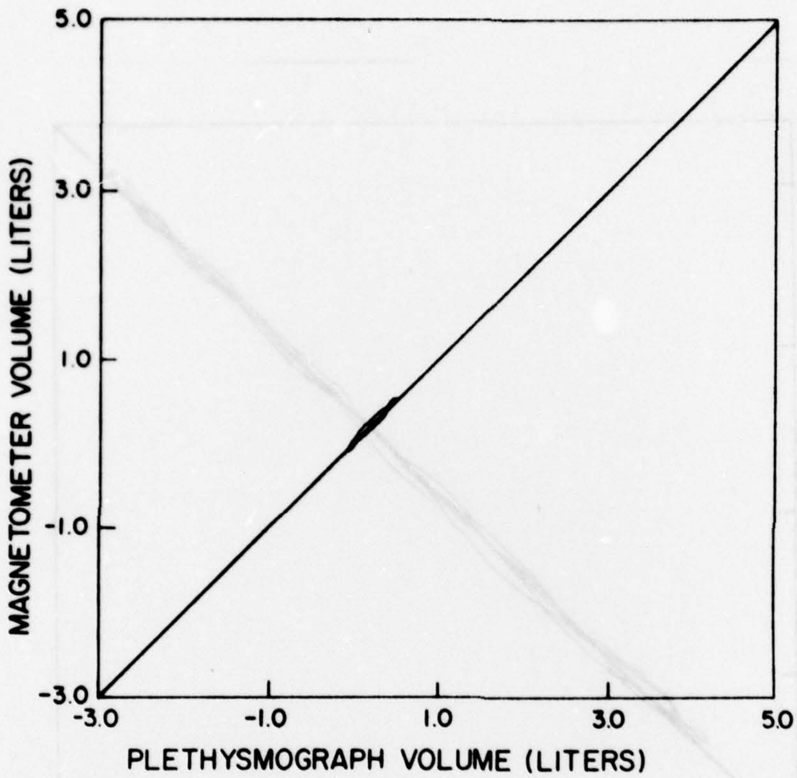

5.0
3.0
1.0
-1.0
-3.0
MAGNETOMETER VOLUME (LITERS)
-3.0
-1.0
1.0
3.0
5.0
PLETHYSMOGRAPH VOLUME (LITERS)

Figure 4A.

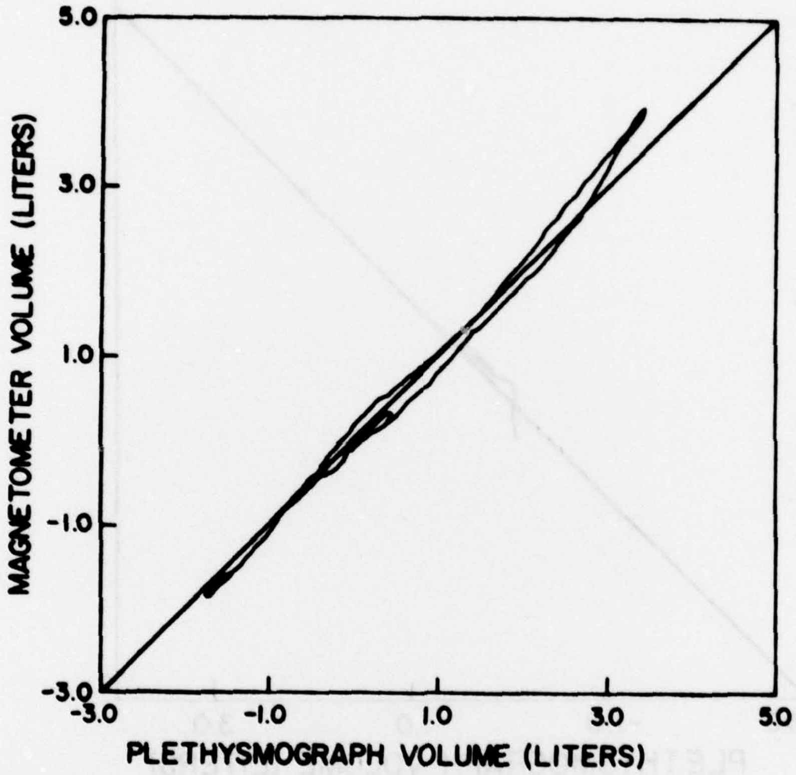

MAGNETOMETER VOLUME (LITERS)
5.0
3.0
1.0
-1.0
-3.0
-3.0
-1.0
1.0
3.0
5.0
PLETHYSMOGRAPH VOLUME (LITERS)

Figure 4B.

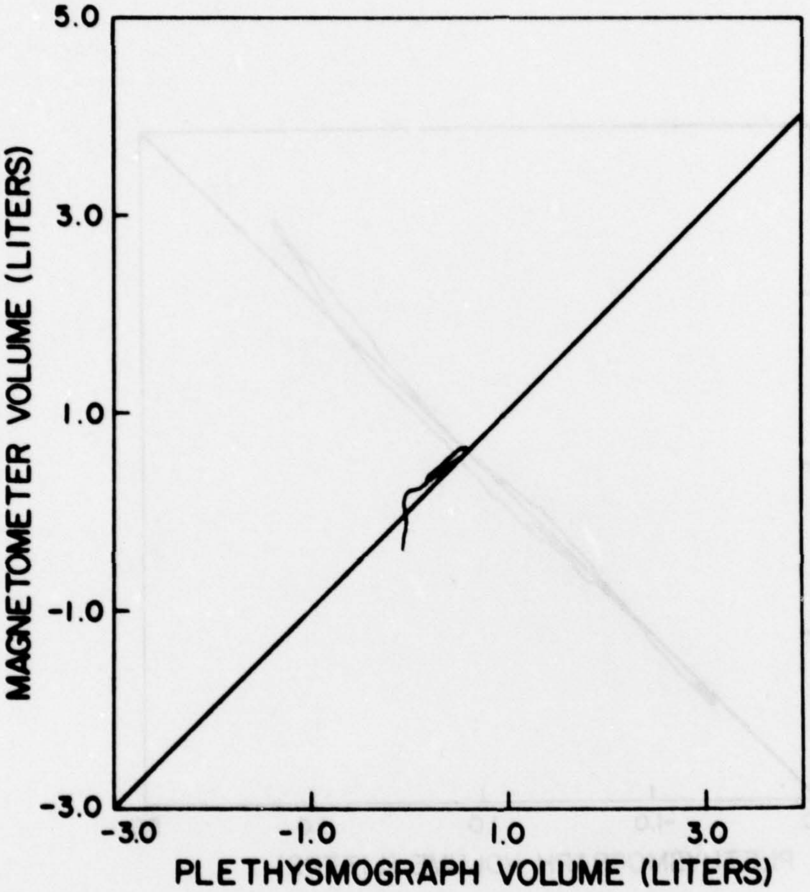

5.0
3.0
1.0
-1.0
-3.0
MAGNETOMETER VOLUME (LITERS)
-3.0
-1.0
1.0
3.0
PLETHYSMOGRAPH VOLUME (LITERS)

Figure 5.

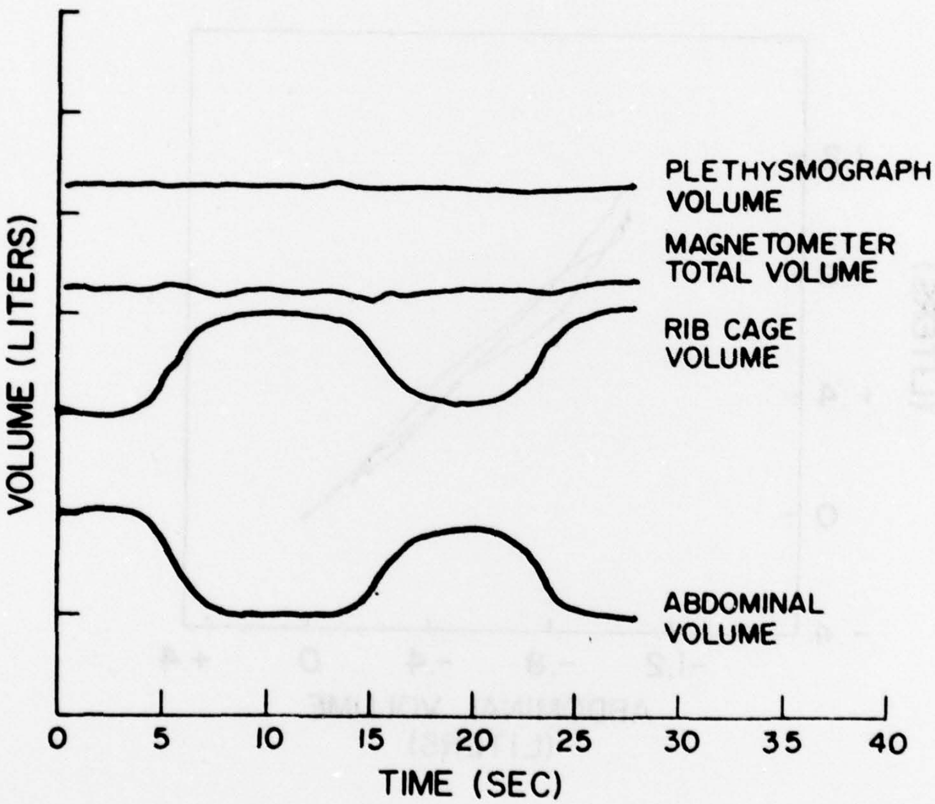

PLETHYSMOGRAPH VOLUME
MAGNETOMETER TOTAL VOLUME
RIB CAGE VOLUME
ABDOMINAL VOLUME
VOLUME (LITERS)
0
5
10
15
20
25
30
35
40
TIME (SEC)

Figure 6A.

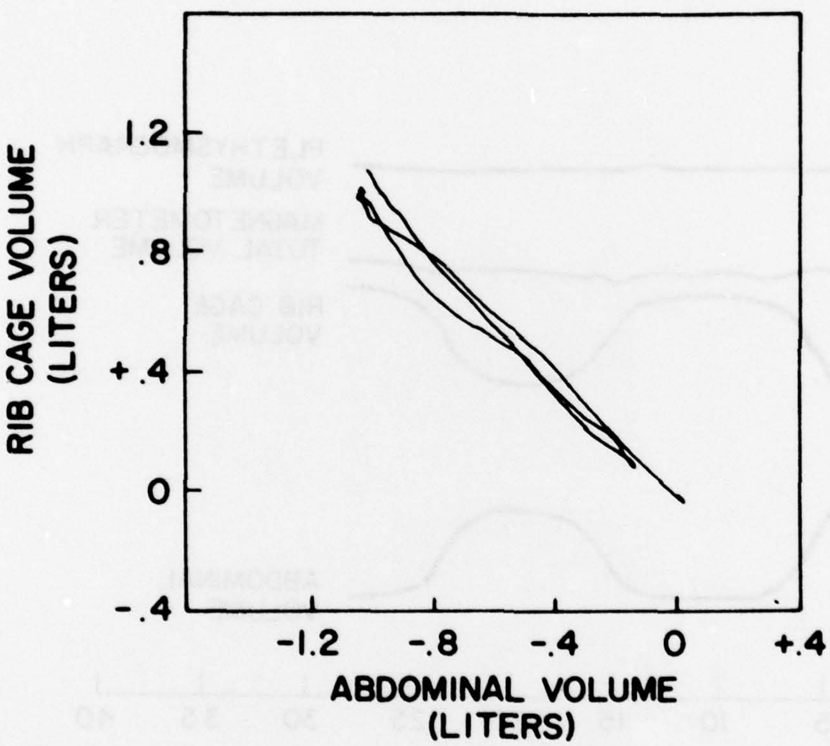

RIB CAGE VOLUME
(LITERS)
1.2
.8
+.4
0
-.4
-1.2
-.8
-.4
0
+.4
ABDOMINAL VOLUME
(LITERS)

Figure 6B.

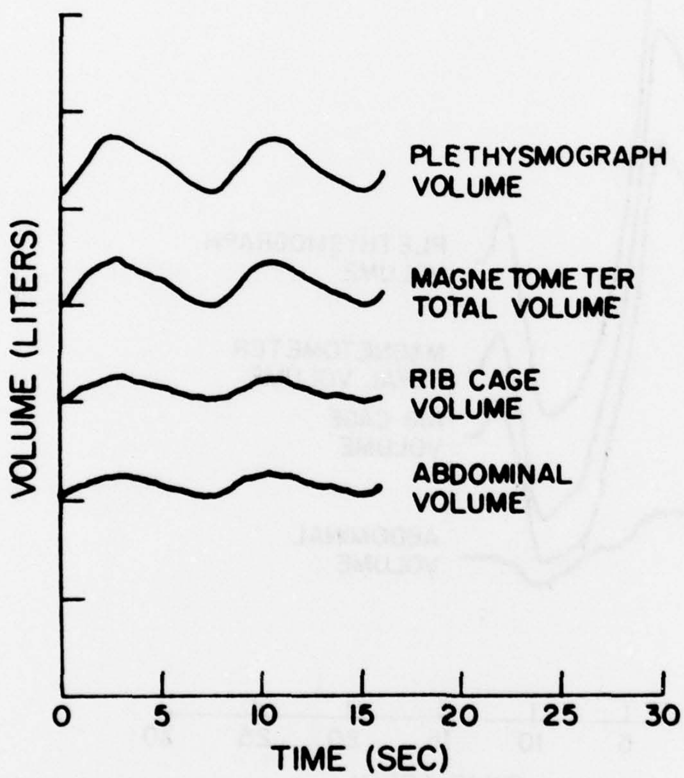

PLETHYSMOGRAPH VOLUME
MAGNETOMETER TOTAL VOLUME
RIB CAGE VOLUME
ABDOMINAL VOLUME
VOLUME (LITERS)
0
5
10
15
20
25
30
TIME (SEC)

Figure 7A.

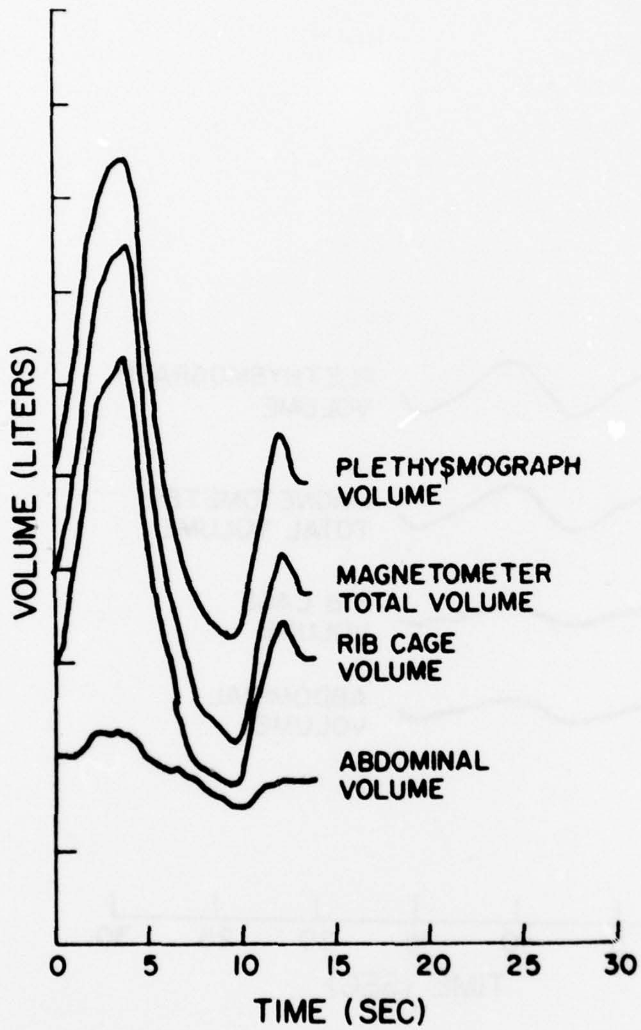

VOLUME (LITERS)
PLETHYSMOGRAPH
VOLUME
MAGNETOMETER
TOTAL VOLUME
RIB CAGE
VOLUME
ABDOMINAL
VOLUME
0
5
10
15
20
25
30
TIME (SEC)

Figure 7B.

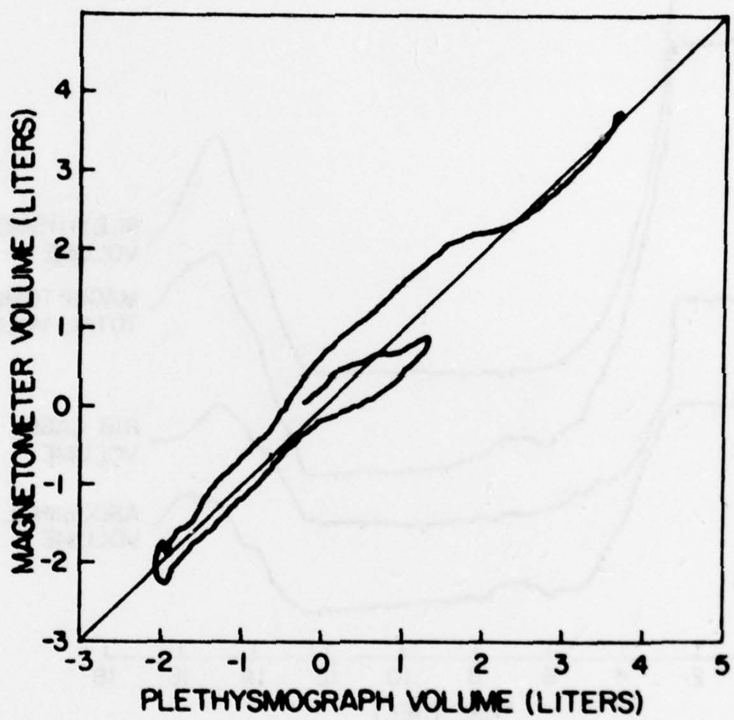
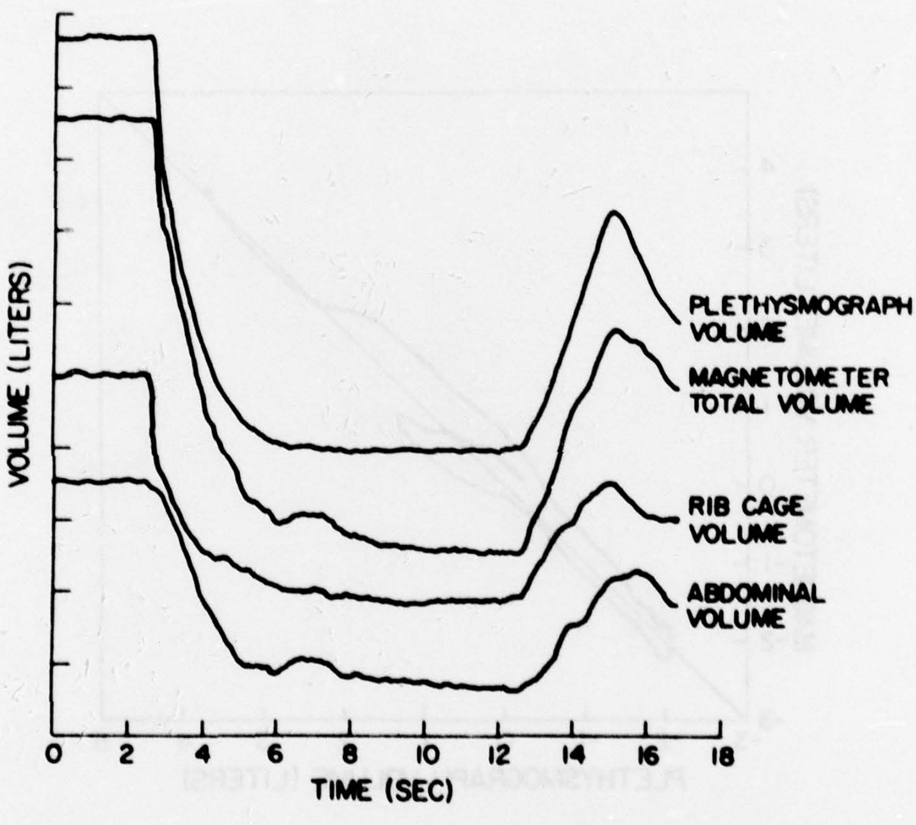

MAGNETOMETER VOLUME (LITERS)
PLETHYSMOGRAPH VOLUME (LITERS)
4
3
2
1
0
-1
-2
-3
-3
-2
-1
0
1
2
3
4
5

Figure 8A.



VOLUME (LITERS)
PLETHYSMOGRAPH VOLUME
MAGNETOMETER TOTAL VOLUME
RIB CAGE VOLUME
ABDOMINAL VOLUME
0 2 4 6 8 10 12 14 16 18
TIME (SEC)

Figure 8B.

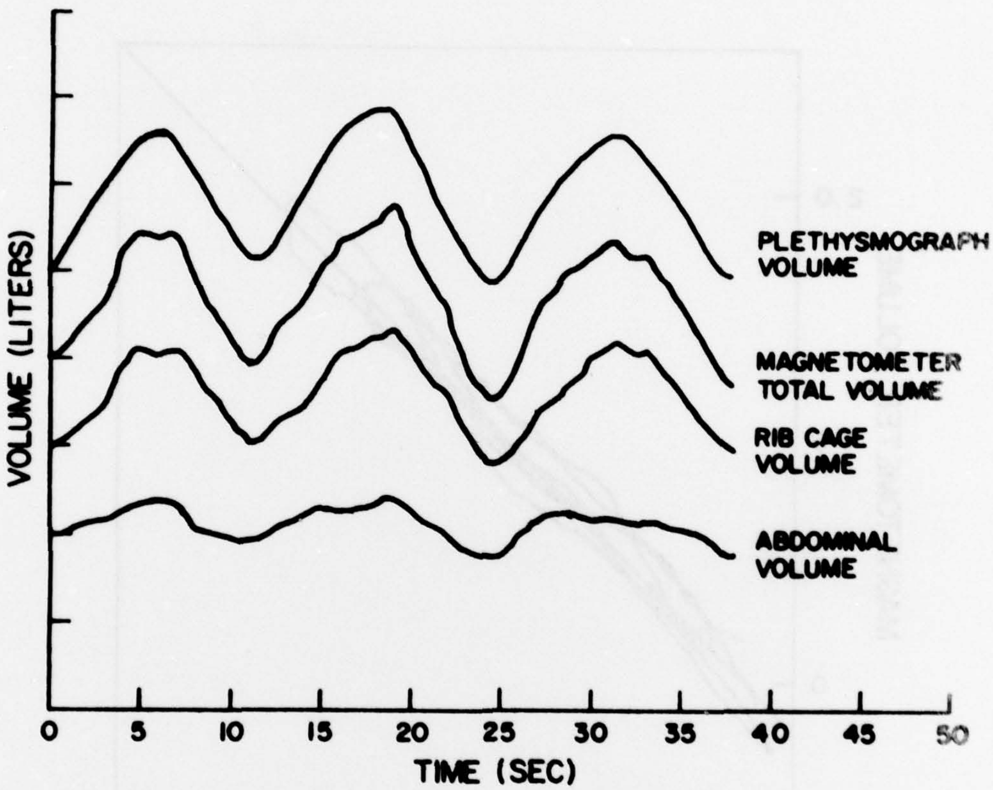
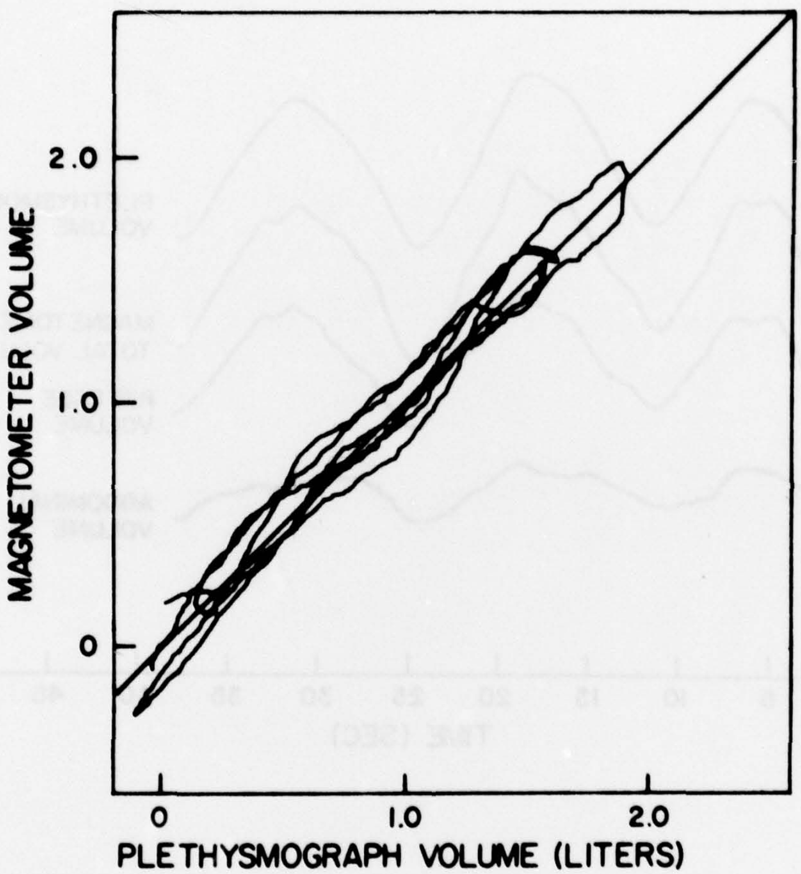

VOLUME (LITERS)
PLETHYSMOGRAPH VOLUME
MAGNETOMETER TOTAL VOLUME
RIB CAGE VOLUME
ABDOMINAL VOLUME
0
5
10
15
20
25
30
35
40
45
50
TIME (SEC)

Figure 9A.



MAGNETOMETER VOLUME
2.0
1.0
0
0
1.0
2.0
PLETHYSMOGRAPH VOLUME (LITERS)

Figure 9B.